SINGLE-LEG CYCLING, AN EVALUATION OF PEDAL POWERS

by

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STATEMENT OF THESIS APPROVAL

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ABSTRACT

Benefits of single-leg cycling may be compromised because single- and double-leg cycling are biomechanically different. Specifically, during normal double-leg (2L) cycling the gravitational forces acting on each leg are essentially balanced by the contralateral limb and thus do not require active leg flexion. Conversely, single-leg (SL) cycling requires active leg flexion. Recently, The Neuromuscular Function Laboratory at the University of Utah devised a counterweighted cycling crank that facilitates SL cycling with similar biomechanics to normal 2L cycling. The purpose of this study was to evaluate SL noncounterweighted pedal powers (SL-0), SL counterweighted pedal powers at 20 and 30 pounds (SL-20 & SL-30), and compare to normal 2L cycling. Eleven trained cyclists (age: 39 ± 7 years, mass: 172 ± 42 pounds, height: 68 ± 3.5 inches) performed SL cycling with their right leg and 2L cycling with both legs. Pedal powers were calculated during each trial using a force measuring pedal and instrumental spatial linkage system. Participants warmed up for 5 minutes and then performed 6 randomized cycling trials while maintaining 90 rpm at 200 watts 2L, and 100 watts SL-0, SL-20, and SL-30. One-way within-subjects ANOVAs indicated significant effects for cycling condition for both extension [$F(1,62) = 26.17$, $p < .01$., partial $\eta^2 = .72$] and flexion [$F(1,73) = 50.68$, $p < .01$., partial $\eta^2 = .835$]. Follow-up pairwise comparisons indicated that 2L cycling generated the most power, 2L-0 the least, with SL-20 & SL-30 falling in between the two extremes, but still significantly different from the low and the high. Within-subjects ANOVAs were conducted using mean-adjusted body weight as a covariate still showed significant differences for power generated under different cycling conditions; also a significant interaction between cycling condition and body weight was observed. The magnitude of power decreased significantly when comparing SL-0 to 2L. However, adding counterweight (SL-20, SL-30) demonstrates evidence that counterweighted cycling brings pedal powers closer to 2L cycling in a SL cycling model. In conclusion, 2L cycling is not the same biomechanical task as SL-0 cycling. Counterweighted single-leg cycling produces similar pedal power to 2L cycling, but the exact application of the counterweight needs further investigation. Properly counterweighted SL cycling may be a beneficial training and rehabilitation modality.

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Finally, I would like to dedicate this manuscript to my wonderful family; my wife Ashley and two children Logan and Bailey. This accomplishment could not be possible without their support and understanding.

CHAPTER 1

INTRODUCTION

During normal double-leg cycling, approximately half the total work and power is produced by each leg. Consequently the oxygen uptake of each leg is likely limited to approximately half of maximal double-leg oxygen consumption (VO_{2max}; Ogita, Stam, Tazawa, Toussaint, & Hollander, 2000). Dynamic exercise such as cycling causes increased oxygen to be delivered to the working skeletal muscles in order to meet increased energy demands. The power that can be sustained by the muscles of each leg during cycling (or any endurance task) is limited by the delivery and extraction of oxygenated blood. The total amount of oxygen that can be metabolized by the working muscles is a function of respiratory capacity. Muscle respiratory capacity is associated with mitochondrial function and is generally considered to be symorphic with oxygen delivery (Rome & Lindstedt, 1997).

Maximal oxygen uptake (VO_{2max}) is the product of maximal cardiac output and arterial-venous oxygen difference, which reflects the body's ability to consume oxygen. Thus, VO_{2max} reflects the combined effects of the cardiovascular system to deliver oxygenated blood and the muscles' ability to extract oxygen from the blood. Submaximal aerobic performance is characterized by a ventilatory threshold (VT), the point at which CO_2 production exceeds O_2 consumption (thereby buffering cytosolic acidosis), and lactate threshold (LT), which represents the accumulation of peripheral metabolic byproducts that occurs as a result of respiratory stress. Thus, increases in VT and LT reflect increases in respiratory capacity, in particular the muscle's ability to utilize oxygen without mitochondrial stress that produces metabolic byproducts. Increases in respiratory capacity are directly attributable to an increase in mitochondrial capacity, and/or capillary density (Hoed, Hesselink, Kranenburg, & Westerterp, 2008). Mitochondrial density limits the muscle's ability to produce adenosine triphosphate (ATP) aerobically through a more elaborate mitochondrial reticulum and function*;* whereas capillary density limits the transport of oxygen from the blood to the muscle cell. Thus, increases in mitochondrial and/or capillary density will contribute to an increase in VT and LT.

Although increasing VT and LT will likely improve submaximal or endurance performance, making those increases can be challenging because the respiratory capacity of the muscle must be stressed (overloaded) in order to produce positive adaptations. The overload principle states physiologic systems will adapt to specific stressors (Wilmore & Costill, 1994). Thus, the respiratory capacity of the working muscles must be aerobically stressed to produce positive adaptations in respiratory capacity. LT values among different populations vary from 50% of VO_{2max} in untrained individuals to 84% of VO_{2max} in trained individuals (Coyle, 1995). The exercise intensities for increasing LT are limited to the top 16 to 33% of VO_{2max}. Therefore, the amount of adaptation that can take place is limited by the exercise intensity at LT; this narrow range limits the overload that can be applied to drive increases in respiratory capacity.

Athletic performance velocity during endurance exercise lasting from a few minutes to 2 hours is determined by the highest steady-state VO_2 that can be tolerated.

The VO_2 steady state maintained during competition is related to the VO_2 at which lactate begins to accumulate in blood (LT; Coyle, Martin, Ehsani, Hagberg, Bloomfeild, Sinacore, & Holloszy, 1983). The ability to increase respiratory capacity results in direct improvements in endurance performance. Although training athletes may be the obvious application for increased respiratory capacity, researchers have reported that congestive heart failure (CHF) patients who can increase their VT through exercise are significantly more likely to survive long enough to receive a heart transplant (Coyle et al., 1983; Gitt, Wasserman, Kilkowski, Kleemann, Kilkowski, Bangert, Schneider, Schwarz, & Senges, 2002), which means training related improvements are truly a life and death matter for these patients. CHF patients have remarkably low VO_{2max} values on the order of 15 ml/kg/min. Due to the central limitations associated with CHF, these patients have a difficult time increasing respiratory capacity, because the work rates they can sustain may be too low to elicit increases in mitochondrial or capillary density.

Clearly, athletes and CHF patients can benefit from increased respiratory capacity, but eliciting those increases can be challenging due to central limitations. An alternate method for overloading peripheral respiratory capacity is through partitioning exercise into smaller muscle masses. Partitioning cycling as a model would be accomplished by cycling with one leg at a time, thus reducing the amount of exercising muscle mass. By reducing exercising muscle mass, single-leg cycling allows the entire supply of oxygenated blood to the working muscles of just one leg. In this way, the entire central capacity would be available for use by the periphery. In single-leg cycling such an increase in oxygen availability should allow for greater work production in each leg, overloading mitochondria, and causing an increase in respiratory capacity.

Therefore, single-leg cycling may allow for greater peripheral muscle respiratory adaptations than double-leg cycling (Klausen, Secher, Clausen, Hartling, & Trap-Jensen, 1982).

Although single-leg cycling should allow for increased delivery of oxygenated blood to the leg, the biomechanics of single-leg cycling impose limitations on exercise volume and intensity. Specifically, during normal double-leg cycling, gravitational and inertial forces acting on each leg are essentially balanced by the contralateral limb. Conversely, single-leg cycling is typically performed by alternatively removing one leg from the pedal and pedaling with the other. With one leg resting, the forces are no longer balanced by the contralateral limb causing a misbalanced motion (Figure 1, A). This single-leg cycling approach requires active leg flexion to lift the leg through the pedal cycle; the muscles that flex the leg are relatively small and more fatigable, which may limit the duration of single-leg cycling. However, if single-leg cycling could be performed with biomechanics that approximate normal bilateral cycling, then theoretically total volume and intensity of exercise training could be increased and greater peripheral adaptations realized.

We have recently devised a counterweight system that allows single-leg cycling to be performed with biomechanics that generally approximate double-leg cycling. A counterweight is attached to the noncycling crank to provide gravity and inertial forces similar to those normally provided by the contralateral limb. This single-leg cycling system is quite simple and may provide a means to create an overloading condition for development of increased respiratory capacity. We have used the single-leg system in previous research (Elmer & Martin, 2010) and have determined the counterweight by approximating a general limb weight (Figure 1, B). Biomechanics may change with different; body masses, abilities, cadences, and work outputs. A range of counterweights at different cadences and work outputs with a variety of body masses has not been explored. Therefore, further investigation of the proper counterweight will provide a more complete and useful model for clinical and performance application of the single-leg system.

Purpose Statement

The purposes of this study were to first quantify pedal power during double-leg cycling. Second, determine pedal power during single-leg cycling without the use of a counterweight. Third, determine pedal power through a range of counterweights in order to determine a counterweight that best follows normal double-leg cycling. Pedal power will be evaluated across a range of power outputs and body masses. The aim of this study was to produce a single-leg cycling model that closely approximated pedal power produced during double-leg cycling. The proposed model could serve as the foundation for further research in: preventative health, clinical rehabilitation, and competitive athletes.

Research Questions

Three research questions were investigated in this study:

1. Does single-leg cycling (noncounterweighted) exhibit similar pedal power compared to normal double-leg cycling?
2. Does a counterweighted single-leg cycling model exhibit similar pedal powers to normal double-leg cycling?

3. Does counterweighted single-leg cycling flexion/extension differ between single-leg counterweighted conditions, compared to double-leg cycling?

Research Hypotheses

The following hypotheses were tested in this investigation:

RQ #1: Double-Leg vs. Single-Leg Pedal Powers

I hypothesize that the pedal power in the extension phase of the pedal stroke will be lower and that pedal power in the flexion phase of a pedal stroke will be greater during noncounterweighted single-leg cycling compared to double-leg cycling.

RQ #2: Single-Leg Model Pedal Power

I hypothesize that a properly counterweighted single-leg cycling pedal powers will be similar to double-leg cycling pedal powers.

RQ #3: Counterweight Selection and Pedal Power Evaluation

I hypothesize that body weight and power production will have an effect on the counterweight applied to produce similar pedal powers between single- and double-leg cycling through flexion and extension phase of the pedal stroke.

Significance

The information obtained in this study will lay the ground work in exploring a new area of training muscular respiratory capacity. The biomechanical analysis will show that indeed single-leg noncounterweighted cycling is not similar to normal double-leg cycling. This study will evaluate and quantify pedal powers through the flexion and extension phases using a model of single-leg cycling, which utilizes a counterweight to simulate normal double-leg cycling. Evaluate and quantify pedal power differences between normal double-leg cycling, single-leg cycling with no counterweight, 20 pound counterweight, and 30 pound counterweight. The evaluation of the aforementioned pedal powers will provide information for future investigations on single-leg cycling.

Delimitations

The following delimitations were made for this investigation:

1. Participants in this study were selected from a convenience sample of trained cyclists currently living in Salt Lake City, UT.
2. Participants were between 19-46 years of age.

Limitations

The following limitations were present for this investigation:

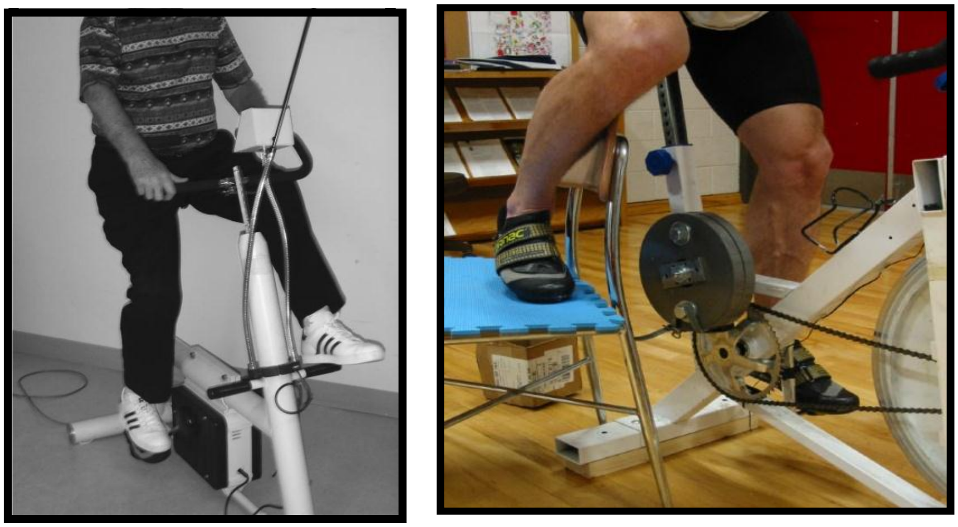
1. Participants may have felt inclined to answer in a socially desirable way to questions of a personal nature.
2. Results from this study were only generalizable to trained cyclists.
3. We aimed to obtain a cohort with a variety of body masses and heights. We may not be able to account for every individual profile.
4. Travel time and distance to the facility may have limited those who would like to participate.

Assumptions

The following assumptions were made for this investigation:

1. The primary investigator was present during all trials and gave the same explanation and encouragement to each participant.

2. The primary investigator followed a specific protocol that included written instructions for completion of the study.
3. Each participant performed to the best of his or her ability during each trial.



A B

Figure 1: Image demonstrating two forms of single-leg cycling. Typical single-leg cycling often used in literature and clinical practice (A). Depiction of the counterweighted system used during single-leg cycling (B).

CHAPTER 2

REVIEW OF LITERATURE

The review of literature consists of the components I feel justify the reasoning for the investigation of single-leg cycling. I will review the current literature on cycling biomechanics, in particular research and reviews on pedal powers during normal upright cycling. Subsequently, I will briefly review the uses of single-leg cycling in research and rehab modalities. Finally, I will describe the previous use of a counterweighted single-leg cycling model in research.

Cycling Biomechanics on Pedal Powers

In the evaluation of pedaling powers, it is generally accepted that the pattern of force applied in a complete pedal cycle begins at zero degrees (the top) and the crank rotation continues clockwise around the 360 degrees of motion. Approximately from zero to 180 degrees is extension (downstroke), and 180 to 360 degrees is flexion (upstroke; Hull & Jorge, 1985). Pedal power can be broken down into tangential or vertical and radial or horizontal components. Pedal power is the dot product of force and velocity. Positive pedal power is produced in extension and negative pedal power is produced in flexion. The production of power creates a sinusoidal wave with more work being done during extension (Figure 2). During maximal cycling, pedal powers have the ability to be manipulated via pedal trajectory and thus alter the mechanism of pedal power (Martin, Lamb, & Brown, 2002) such mechanisms will not be included in the scope of this paper. In a review article by Kautz and Neptune (2002), it was described that net [pedal] power produced through the crank cycle of normal bilateral cycling is the redistribution of segmental energy by muscle forces that link changes in total mechanical energy of the legs to external work production. The production of external work when segmental energy is redistributed by muscle force explains the result that external work exceeds the net muscle work for a large portion of the crank cycle. Work performed through the crank cycle is composed of energy increases and decreases of the leg and are the result of redistribution of segmental energy by muscle forces. The energy decreases resulting from the deceleration of the legs generates a pedal force tangential to the crank (Redfield & Hull, 1986). Through the redistribution of segmental masses, counter-torque (work done in flexion) actually produces little negative work. Essentially during normal double-leg cycling, gravitational and inertial forces acting on each leg are balanced by the contralateral limb. I hypothesized the evaluation of normal bilateral cycling would demonstrate balanced acceleration and deceleration of leg segmental power production at the pedal.

Implications to Single-leg Cycling

Although normal double-leg cycling demonstrates cooperative effort between both legs counteracting flexion, single-leg cycling poses a counter-torque conundrum. To my knowledge no researcher has addressed specifically the biomechanical implications of single-leg cycling. An attempt to evaluate forces of single-leg cycling was made by Sargeant and Davies (1977), but no other researcher has further evaluated single-leg cycling biomechanics. In the study by Sargeant and Davies (1977), a cycle ergometer was used that used a fixed gear and flywheel. The inertia created by the flywheel and fixed gear was justified to have made up the nonsignificant difference between double- and single-leg cycling. The inertia and fixed gear would not completely eliminate the counter-torque effect of the missing limb and thus may significantly alter mechanical efficiency and metabolic cost. Single-leg cycling is typically performed by alternatively removing one leg from the pedal and pedaling with the other. When a limb is removed from the pedaling cycle, it removes the counter-torque described by Kautz and Neptune (2002). With forces no longer balanced, single-leg cycling requires active leg flexion to lift the leg through a complete pedal cycle, which in turn causes decreased power production during extension.

For this reason, I propose that single-leg cycling becomes a different biomechanical task in relationship to normal bilateral cycling.

Single-Leg Research to Date

Cycling is often used as a modality in research due to the ability of the investigator to manipulate and measure many physiological variables while consistently controlling the working environment. Cycling is also a dynamic task that is contained in a closed circuit where data can be collected rather easily. In this section, I will briefly review studies that have utilized a single-leg cycling model in research to date.

Adaptations to Single-leg Training

The majority of studies using a single-leg cycling model did so to compare results to double-leg cycling. Researchers used one leg in comparison with two legs in measuring some physiological mechanism or training adaptations. Initial research was focused on determining general physiological differences in oxygen uptake, arterial pressures, lactate production, and hemodynamics between single- or double-leg cycling (Freyschuss & Strandell, 1968). It was observed that all values measured were significantly higher for single-leg cycling given oxygen uptake, and in conclusion was thought to be due to a higher sympathetic outflow. It should be noted that all measurements were taken at the same workload whether using one or two legs during cycling. Freyschuss and Strandell (1968) attempted to make up for active leg flexion by adding two to four springs to aid in the flexion phase of the crank cycle.

In the early 70s Gleser (1973) and Davies and Sargeant (1974) were specifically interested in evaluating hemodynamics and hyper/hypoxia (with 45% oxygen and 12% oxygen respectively). Gleser (1973) performed a training study, whereas Davies and Sargeant (1974) only observed differences; both were interested in both submaximal and maximal values between single- and double-leg cycling. Both studies used the same workloads between the single- and double-leg cycling trials. Gleser (1973) tried to correct for active leg flexion by springs and coordinated pedaling between two participants whereas Davies and Sargeant (1974) did nothing to correct for the flexion phase. Both studies evaluated the metabolic components and indicated that only approximately 70% of aerobic power could be reached, while consuming relatively more oxygen. They suggested countering views of limitation; Gleser (1973) attributed the lower VO_{2max} to limitations of the peripheral capacity whereas Davies and Sargeant (1974) to limitations of cardiac output or central capacity.

The conflicting views most likely lead to follow-up studies by Davies and colleagues (Davies & Sargeant, 1975; Futoshi, Roelof, Haruhik, Huub, & Hollander, 2000; Klausen, Secher, Clausen, Hartling, & Trap-Jensen, 1982; Magnusson, Kaijser, Isberg, & Saltin (1994); Stamford, Weltman, & Fulco, 1978). All the research evaluated differences in oxygen consumption, ventilation rate, heart rate, and cardiac output. The methods varied between investigators for workloads administered from using a percentage of single-leg and double-leg oxygen consumption, a set heart rate, or set wattage production. None of the studies evaluated the possible effects of differences in biomechanics between single- and double-leg cycling, and none of the studies actively tried to account for the flexion phase in single-leg cycling. The results were somewhat similar in that all concurred there appears upregulatory capacity in the periphery through training single-leg. It seems that although training single-leg may increase the ability to consume additional oxygen in that particular limb and produce additional work, when combined with double-leg cycling almost no global change in oxygen consumption is observed. In general, this is concluded that in the instance of single-leg cycling there is no limitations of central delivery, and additional availability of blood flow, oxygen, and metabolic byproducts allows for upregulation of respiratory capacity. But in double-leg cycling there may be a central limiting factor, which when single-leg training has taken place is not able to supply sufficient blood flow, oxygen, and metabolic byproducts. The VO_{2peak} observed for single-leg cycling was approximately 80-85% of that for double-leg cycling in the Davies and colleagues (Davies & Sargeant, 1975; Futoshi et al., 2000; Klausen et al., 1982; Stamford et al., 1978) research. All suggest there is further research that needs to be performed to clarify in more detail cardiac output, blood flow, and metabolic responses in single- and double-leg cycling.

Ray (1993, 1999) focused on evaluating the neurological components of short interval training and long endurance training on muscle sympathetic nerve activity. Ray (1993, 1999) reported training can have some effect on neural response during and after exercise bouts. Specifically, muscle sympathetic nerve activity is decreased during the early stages of upright single-leg cycling and elevated during recovery when heart rate and mean arterial pressure are at control levels. The mechanisms are still unknown for the nervous responses, and illicit further investigation.

Disease Populations

Results from the previous studies suggest increased peripheral adaptation may be possible through single-leg cycling. Researchers evaluated possible increase respiratory capacity for disorders/diseases that limit its recipients by central delivery such as CHF and chronic obstructive pulmonary disease (COPD). A recent study done by Dolmage and Goldstein (2008) evaluated single-leg cycling on COPD patients. Methods included both a double-leg and single-leg cycling group that performed intervals of training, both groups performed only 30 minutes, 3 days per week for 7 weeks. Training single-leg cycling reduced the total central metabolic demand and improved aerobic capacity compared with conventional two-legged training in patients with COPD. An additional study has also been done using a single-leg model in evaluating monoparesis patients (White & Dressendorfer, 2005) where the investigators evaluated the maximal oxygen uptake in both legs in a multiple scerolosis patient displaying left monoparesis. The investigation concluded that the ability to deliver oxygen to the paretic limb was not limited, but the strength and/or muscular oxidative capacity was considered to be the limiting factor.

I was unable to find a study where researchers used a single-leg model to investigate possible CHF training adaptations. There is sufficient evidence to suggest the ability to increase peripheral capacity through a single-leg cycling modality. As discussed in the introduction, respiratory capacity for a heart patient is vital for both quality of life and life expectancy.

Counterweighted Single-leg Cycling

To date The Neuromuscular Function Laboratory at the University of Utah has used the counterweighted single-leg cycling model in a handful of research projects specifically Elmer and Martin (2010) dynamically quantifying neuromuscular function after eccentric muscle damage and during recovery. Elmer and Martin (2010) extensively used counterweighted single-leg cycling in both maximal and submaximal evaluations of work, rate of perceived exertion (RPE), and neuromuscular responses. Miller (2009) presented on bilateral deficit in maximal power production utilizing a counterweighted single-leg cycling model, and reported no power differences between right and left legs, but observed changes in maximal oxygen uptake.

Martin recently collaborated a further evaluation of oxidative markers and metabolic changes (Abbiss, Laursen, Karagounis, Peiffer, Martin, Hawley, Fatehee, & Martin, 2010) using the counterweighted single-leg cycling model. The research evaluated the training effect of a 3-week high-intensity interval training design on single-leg cycling compared to normal double-leg cycling. The investigators found the trained cyclists were able to produce more work at a reduced central demand and RPE during single-leg cycling and elicited a significantly greater oxidative and metabolic potential in skeletal muscle. This conclusion further supports the importance of the single-leg cycling model to research.

In the current investigation, I have attempted to provide a validation of the single-leg cycling model used in previous investigations. The single-leg cycling model developed in this investigation provided a specific method of dynamically evaluating differences between single- and double-leg cycling. The validation of a biomechanically correct model may lead to a reevaluation of initial investigations on possible physiological differences and training adaptations to single-leg cycling.

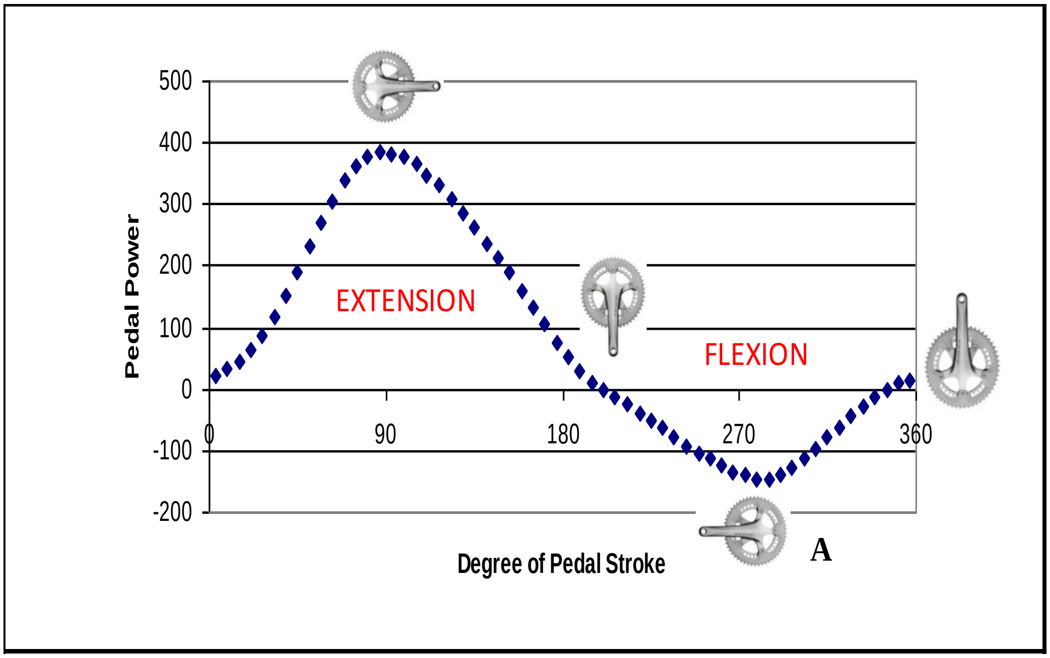


Figure 2: Graph depicting measured pedal power of 200 watts on the right foot. The graph contains a crank (A) to understand rotation of crank through 0-360^{o}. The sinusoidal curve contains an area of extension and flexion.

CHAPTER 3

METHODS

In this chapter, I will explain the methodology used in this investigation. Particularly, I will describe the participants who volunteered for the study, experimental procedures, instrumentation, and research design and analyses.

Participants

The methods to be used in this study have been reviewed by the Institutional Review Board at the University of Utah. A convenience sample of 11 trained cyclists was recruited for participation in the study. Sample size for this study was estimated from the repeated measures tables provided by Green (1990) and Stevens (2002). Twenty sampling units (n = 10/treatment group) were required assuming a large effect size and power of .8 (alpha = .05). Eleven participants (n = 11/treatment group; N = 22) were recruited, which allowed for one participant dropout. Trained cyclists were defined as those who participate in cycling as a form of exercise for at least 5 hours per week. The number of participants recruited for this study was more than double that used in other studies analyzing single-leg cycling (Klausen et al., 1982; Ogita et al., 2000).

In order to assess a wide range of body masses, I tried to find a convenience sample containing a wide range of body types (height, mass, sex). A wide range of variability was needed to explore counter-weighted single-leg cycling, as I believe height and weight may play a role in the counter-weight needed for optimal single-leg cycling biomechanics. Verbal and written explanations of the study were provided to each of the participants so they were fully aware of the time commitment and level of exercise intensity that this study required. An informed consent form was completed, dated, and signed by each participant. Individuals with any lower extremity injuries were excluded from this study in order to eliminate the possibility of further injury. To ensure consistency in metabolic responses women were tested 10 days following menstruation.

Investigation Timeline

This investigation took place over one week. Participants were required to report to The Neuromuscular Function Laboratory on one occasion. The laboratory visit consumed approximately 90 minutes of their time. Figure 3 provides an overview of the timeline during the laboratory visit.

Experimental Procedures

Consent and General Biometrics

Each participant reported to The Neuromuscular Function Laboratory (HPER West 122) on one occasion. The purpose of the visit was to deliver verbal and written explanations, collect informed consent forms, and collect experimental data. Measurements were collected (age, height, mass, foot length, and thigh length) along with a brief questionnaire regarding history (cycling experience).

Pedal Power Collection

Participants warmed up for 5 minutes and then performed six randomized cycling trials of 105 seconds (Figure 4). During each 105 second trial, the participant maintained a specified pedaling rate (90 rpm) while the cycle ergometer power requirement increased in load every 15 seconds. The first load in each bout was constant for 30 seconds to provide enough time for the participant to get accustomed to the load and cadence. The loads for double-leg cycling were 50, 100, 150, 200, 250, and 300 watts and the loads for single-leg cycling were 25, 50, 75, 100, 125, and 150 watts. The participant then received approximately 3 minutes rest between each trial.

As indicated in Figure 4, participants performed double-leg cycling trials at pedaling rate of 90 rpm. Subsequently, participants performed six single-leg cycling trials at 90 rpm pedaling rate and with counterweight masses of 0, 10, 20, 25, and 30 pounds. Although the conditions in Figure 4 are numbered, the order of implementation was randomized.

Complete data were collected as described in the previous paragraphs. Due to inconsistencies between participants, trials, and pedal powers, the analysis could not effectively be run as initially anticipated. The data collection produced an immense amount of sheer raw data; I decided to limit the data used for analysis in order to effectively utilize a portion of the data set. I limited the data to the most consistent condition between subjects, which I anticipated would best demonstrate the research questions and hypothesis. I analyzed statistically only one condition; double-leg, single-leg no counterweight, single-leg 20 pound, and single-leg 30 pound, at 90 rpm and 200 watts double-leg and 100 watts single-leg. Further dialogue on possible reasons for errors in data collection is contained in more detail in Chapter 4.

Instrumentation

Cycle Ergometer

A modified Monark (Vansbro, Sweden) cycle ergometer frame and Velotron flywheel system (RacerMate LLC) were used to construct a cycle ergometer (Figure 5, A,B). The ergometer is fixed to the floor and has been fitted with bicycle-racing handlebars, cranks, pedals, and seat. Participants completed the cycling tests on the modified cycle ergometer, which was set up to match their preferred cycling position. Additionally, participants wore cycling specific shoes with spring-load cleats that lock onto the pedals (Figure 5, C). The Velotron is a computer controlled electronic bicycle ergometer. The Veletron's mechanical design provides accuracy and repeatability (McDaniel, Subudhi, & Martin, 2005). The system uses a patented eddy current brake built around a 55 pound, 25 inch diameter flywheel with an internal freewheel and uses a fixed ratio chain drive. Pedal powers were measured during each of the 12 cycling trials.

Force Pedal

The right pedal (Figure 5, C) of the cycle ergometer was equipped with two 3-component piezoelectric force transducers (Kistler 9251: Kistler USA, Amherst, NY, USA), and the right pedal and crank were equipped with digital position encoders (U.S. Digital model S5S-1024: Vancouver, Washington, USA). Normal and tangential pedal forces and pedal and crank positions were recorded for 105 sec at 120 Hz using Bioware Software Version 3.0 (Kistler USA, Amherst, NY, USA). The normal and tangential pedal forces were resolved into vertical and horizontal components using the pedal and crank position data. Pedal power was calculated as the dot product of pedal force and linear velocity.

Anterior superior iliac spine (ASIS) positions were recorded with an instrumented spatial linkage (ISL); Figure 6). Details of the ISL are described by Martin, Elmer, Horscroft, Brown, and Shultz (2007). Briefly, the ISL consisted of a ground anchored base, two aluminum segments, bearings, and digital encoders. The end of the ISL segment was mounted to a threaded connector that was centered on the participant's ASIS and held in place with belt tension. Position data from the digital encoders were also recorded using Bioware software. The ISL used served as a cost-effective, accurate, and valid measure for two-dimensional kinematic data within the typical range of motions for cycling (Martin et al., 2007).

Prior to the protocol, pedal, crank center, and individual greater trochanter and ASIS positions were determined by a one second static data collection of each participant using the ISL. During the exercise protocol, ASIS and pedal and crank position coordinates were measured, which allowed sagittal plane leg segment positions to be determined. More specifically, the coordinates of the ASIS during the exercise protocol and the known distance between the ASIS and the greater trochanter from the static shot allowed for the position of the great trochanter to be inferred throughout the pedal cycle. In addition, crank angle, pedal angle, and the angle created between the pedal and the lateral malleolus allowed for the coordinates of the lateral malleolus to be determined throughout the pedal cycle. With the coordinates of the lateral malleolus and greater trochanter and the known lengths of the foot, shank, and thigh a triangle was formed with known sides. Using the law of cosines ($c^2 = a^2 + b^2 – 2ab\cos C$) the joint angles at the knee and hip were determined. From these data, it was possible to calculate joint angular velocities and accelerations at the ankle, knee, and hip. Linear and angular velocities and accelerations of the limb segments were determined by finite differentiation of position data with respect to time. Position data were filtered using a fourth order zero lag Butterworth Filter and a cutoff frequency was determined based on the recommendations provided by Winter (2005).

Segmental masses, moments of inertia, and location of centers of mass were estimated using the regression equations of de Leva (1996). Sagittal plane joint reaction forces and net joint moments at the ankle, knee, and hip were determined by using inverse dynamic techniques (Elftman, 1939). Joint powers were calculated as the product of net joint torques and joint angular velocities. Power transferred across the hip joint was calculated as the product of the hip joint reaction force and linear velocity. Calculated values for ankle, knee, hip joint, and hip transfer power were averaged over all the complete pedal cycles within the data collection interval.

Research Design and Analysis

RQ #1: Double-Leg vs. Single-Leg Pedal Powers

In order to compare normal double-leg cycling with nonweighted single-leg cycling, a Repeated Measures ANOVA was performed for flexion and extension.

RQ #2: Single-Leg Model Pedal Power

In order to compare the single-leg model to the normal double-leg cycling, noncounterweighted cycling, and the counterweighted (20 and 30 pound) conditions a Repeated Measures ANOVA was performed. The goal was to evaluate the counterweighted single-leg model utilizing counterweights (20 and 30 pound) compared to normal double-leg cycling. The flexion and extension phases of the pedal stroke were individually analyzed for each condition.

After running the Repeated Measures ANOVA, I found a large variance, and thought there may be confounding variables. I ran additional ANCOVA's for flexion and extension phases taking into account possible confounding variables of age, sex, and weight. I discovered that weight was the only confounding variable, and ran follow-up analysis appropriately.

RQ #3: Counterweight Selection and Pedal Power Evaluation

In order to demonstrate the relationship between normal double-leg cycling and counterweighted conditions, a Repeated ANOVA was run.

Participant Time-Table for Laboratory Visit

Aprox Time (min)	Reps	Activity
20		*Read and Sign Consent*
15		*Biometric Measurments Obtained*
5	1	*Warm Up*
2	1	*Double-Leg Trial*
2	6	*Randomized Single-leg Trials*
3	7	*3 min Recovery Between Each Trial*
5	1	*Cool Down*
92		Approximate Total Time

Figure 3: Diagram of participant itinerary during the laboratory visit.

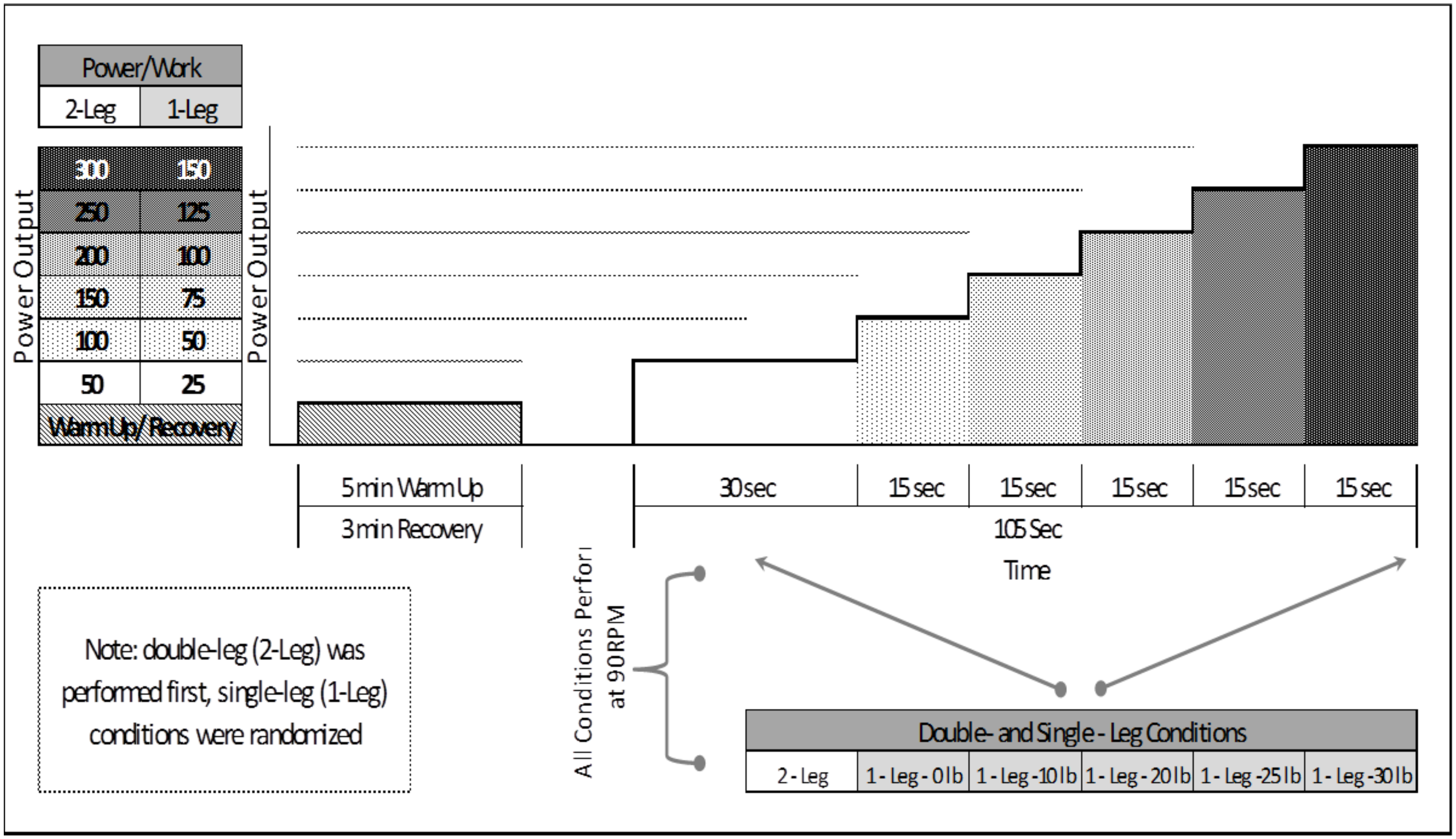


Figure 4: Diagram depicting the complete protocol of the study. Including both single- and double- cycling trials. Double-leg cycling trials were always performed first and all single-leg cycling conditions were randomized.

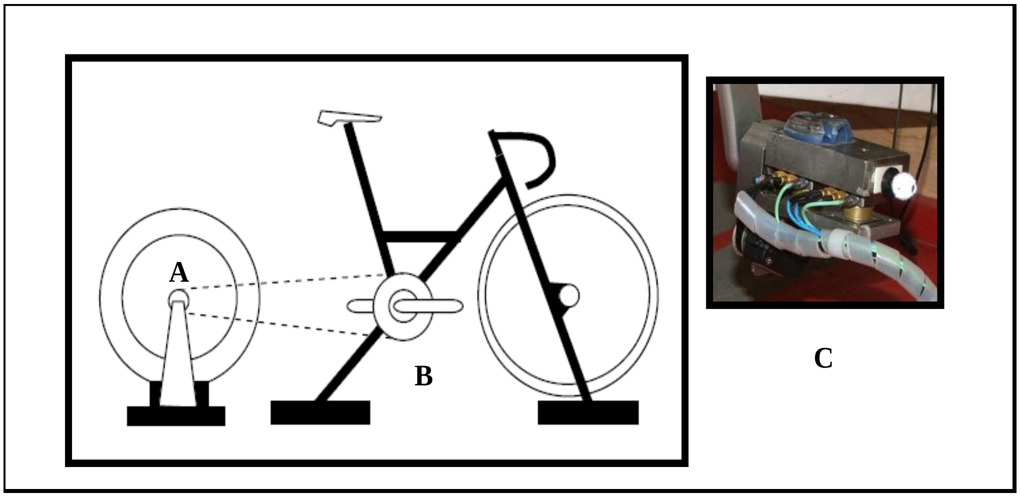


Figure 5: Depiction of the bicycle ergometer used to collect data. The Velotron flywheel (A) was attached to the Monarck bicycle ergometer (B). The force pedal described (C) was also the locking pedal used by participants.

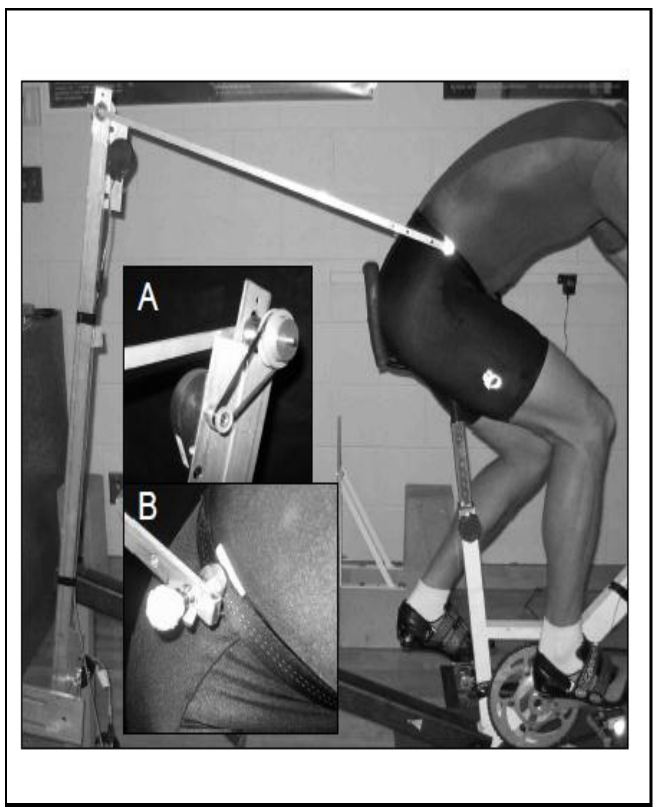


Figure 6: Two-segment instrumented special linkage (ISL). Inset images show detail encoder drive system (A) and the ISL-human interface (B).

CHAPTER 4

RESULTS AND DISCUSSSION

In this chapter, I first present the results of the data analyses and later discuss the results with respect to my research questions and include discussion for each section.

Analysis

The complete sample consisted of 11 participants, 7 males (n = 7) and 4 female (n = 4). All participants cycled a minimum of 5 hours per week and had a variety of cycling backgrounds, which included competitive racing, recreational riding, and street commuting. Overall and subgroup demographics are presented in Table 1. Participants completed all aspects of the study.

Results

A one-way within-subjects ANOVA was conducted with the factor being cycling mode and the dependent variable being power during the extension phase of cycling. The means and standard deviations for power are presented in Table 2. Normality was verified by examining histograms. Mauchly's Test of Sphericity was significant so results were interpreted using the Greenhouse-Geisser adjustment. The results for the ANOVA indicated a significant effect for condition, $F(1.62) = 26.17$, $p < .01$., partial $\eta^2 = .72$.

Follow-up pairwise comparisons were conducted to identify differences between conditions in cycling power during extension. A significance level of .01 was used to adjust for multiple comparisons. Significant differences were seen between the single-leg cycling conditions and the double-leg cycling condition. Significant differences were also seen between the single-leg cycling conditions except between the counterweighted conditions of 20 and 30 pounds. Power decreased significantly when comparing the single-leg cycling conditions to double-leg cycling; however, adding counterweight to the opposite pedal did increase power in the single-leg cycling conditions. The more counterweight that was added, the higher the resulting single-leg power that was observed.

A second within-subjects ANOVA was conducted using mean-adjusted body weight as a covariate. Mauchly's Test of Sphericity was not significant, so results were interpreted using the sphericity assumed results. A significant interaction between cycling condition and body weight was observed, $F(3) = 4.36$, $p = .013$, partial $\eta^2 = .326$.

To evaluate the significant interaction, cyclists were divided into groups according to body weight. Three groups were created: the low group had body weights more than one standard deviation lower than the mean weight, the second group had body weights within one standard deviation above or below the mean body weight, and the high group had body weights greater than one standard deviation above the mean body weight. The within-subjects ANOVA was repeated using the newly created group variable as a between-subjects factor. The main effect of group was not significant. This was most likely due to the small sample size. However, a graph showing power generated under the different conditions divided by groups highlighted the trend responsible for the significant interaction seen originally when body weight was used as a covariate. As expected those with greater body weight generated higher power except in the single-leg no counterweight condition where all groups fell to almost the same level. It is also observed that as counterweight increases in the single-leg cycling conditions, power generated also tends to increase toward double-leg levels (Figure 7).

A one-way within-subjects ANOVA was conducted with the factor being cycling condition and the dependent variable being power during the flexion phase of cycling. The means and standard deviations for power are presented in Table 2. Normality was verified by examining histograms. Mauchly's Test of Sphericity was significant so results were interpreted using the Greenhouse-Geisser adjustment. The results for the ANOVA indicated a significant effect for condition, $F(1.73) = 50.68$, $p < .01$., partial $\eta^2 = .835$.

Follow-up pairwise comparisons were conducted to identify differences between conditions in cycling power during flexion. A significance level of .01 was used to adjust for multiple comparisons. Significant differences were seen between the single-leg cycling conditions and the double-leg condition. Significant differences were also seen between the single-leg cycling conditions except between the 20 and the 30 pounds condition. The magnitude of power decreased significantly when comparing the single-leg conditions to the double-leg condition; however adding counterweight to the opposite pedal did increase power in the single-leg cycling conditions. The more counterweight was added, the higher the resulting single-leg power observed.

A second within-subjects ANOVA was conducted using mean-adjusted body weight as a covariate. Mauchly's Test of Sphericity was not significant, so results were interpreted using the sphericity assumed results. A significant interaction between cycling condition and body weight was observed, F(3) = 12.20, $p < .01$, partial $\eta^2 = .575$.

To evaluate the significant interaction, cyclists were divided into groups according to body weight. Three groups were created: the low group had body weights more than one standard deviation lower than the mean weight, the second group had body weights within one standard deviation above or below the mean body weight, and the high group had body weights greater than one standard deviation above the mean body weight. The within-subjects ANOVA was repeated using the newly created group variable as a between-subjects factor. The interaction effect was significant, F(6) = 4.46, $p < .01$, partial $\eta^2 = .527$. The main effect of condition was significant, F(3) = 68.30, $p < .01$, partial $\eta^2 = .895$. The main effect of group was significant F(2) = 10.19, $p < .01$, partial $\eta^2 = .718$. A plot showing power generated under the different conditions divided by groups highlighted the trend responsible for the significant interaction seen originally when body weight was used as a covariate. As expected, those with greater body weight generated higher power except in the single-leg no counterweight condition, where all groups fell to almost the same level. It is also observed that as counterweight increases in the single-leg cycling conditions, power generated also tends to increase toward double-leg levels, except in the highest weight group where there was a decrease observed in power when counterweight increased from 20 to 30 pounds (Figure 8).

Discussion

The purpose of the present investigation was to evaluate pedal power during cycling, specifically normal double-leg cycling with noncounterweighted single-leg cycling and a counterweighted single-leg cycling model that would enable similar pedal powers to double-leg cycling.

Biomechanical Implications of Double-leg and Noncounterweighted Single-leg Cycling

Although the original hopes for the analysis of data collection was to provide a means to prescribe a counterweight for single-leg cycling based on sex, body mass, and specificity of population, I was unable to show a complete picture. I collected a huge amount of raw data and fortunately was able to use a portion of consistent data between participants to enable conclusions to be drawn. My findings support that of Kautz and Neptune (2002) with pedal power production through a crank cycle in double-leg cycling. From the data, I conclude that normal double-leg cycling pedal powers are significantly different from noncounterweighted single-leg cycling pedal powers in both extension and flexion. It can then be concluded that normal double-leg cycling is a different biomechanical task than noncounterweighted single-leg cycling.

Further investigation of cycling power production through the crank cycle have been produced to break down joint specific (ankle, knee, and hip) kinematics during cycling (Ericson, 1986; Ericson, Bratt, Nisell, Nemeth, Ekholm, 1986; Gregersen & Hull, 2003; Martin & Brown, 2009, 2007). The knowledge of the joint kinematics power production and electromyogram (EMG) can be combined to evaluate muscular contributions to the crank cycle (Ericson, 1988a, 1988b; Hug, Decherchi, Marqueste, & Jammes, 2004; Prilutsky & Gregory, 2000) to further identify muscular components of the crank cycle. Through deduction I may propose that if pedal powers are significantly different in both extension and flexion, we can also assume the power generated by the ankle, knee, and hip may also be different. If the joint kinematics are not similar to what has been identified in double-leg cycling conditions, one could also assume muscle groups involved in extension and flexion would also be different. Investigators have also linked joint kinematics and muscular contribution to metabolic cost (Hug et al., 2004; McDaniel, Durstine, Hand & Martin, 2002; McDaniel et al., 2005). If the metabolic and mechanical changes that occur in noncounterweighted single-leg cycling are significant; all the previous data comparing mechanical efficiency, metabolic cost, and other physiological measures may need to be reinvestigated. Further analysis should be performed to evaluate changes in joint torques and powers in the ankle, knee, and hip during noncounterweighted single-leg cycling and compare them to the vast (Davies & Sargeant, 1975; Futoshi et al., 2000; Klaus et al., 1982; Stamford et al., 1978) literature evaluating differences between single- and double-leg cycling.

Counterweighted Single-leg Cycling Biomechanics

Counterweighted single-leg cycling did not produce the results anticipated. In extensive practice sessions at the laboratory, I was able to use the counterweighted single-leg model and produce a nonsignificant difference between double-leg cycling and counterweighted single-leg cycling. The pedal powers were so similar in fact the original plan for the analysis of the data was to create a regression analysis to determine the correct counterweight for specific populations; specifically sex, body mass, and application (clinical/rehab/performance). I also assumed from my practice sessions prior to collecting data on participants that the counterweighted single-leg model was refined enough that any naive participant would be able to perform counterweighted single-leg cycling using the corrected biomechanical via counterweights. I discovered during the preliminary analysis of data that my participants unintentionally would actively flex the working leg even with a counterweight applied. I feel that through a week of practice sessions the active leg flexion problem could have been diminished or eliminated (Martin, Diedrich, & Coyle, 2000).

The data indicated that counterweighted single-leg cycling is significantly different from normal double-leg cycling as well as significantly different from noncounterweighted single-leg cycling. When I split up the groups into three weight classes, the relationship between the counterweight and individual become more evident (Figure 7, 8). The results would seem to imply utilizing a wider range of counterweights combined with practice sessions would present anticipated results derived from pilot data. The data presented here provides enough evidence that it could be said counterweighted single-leg cycling brings pedal powers closer, but still lower to normal double-leg cycling. Further investigations with this counterweighted single-leg cycling model should be evaluated to complete these findings.

Metabolic Cost Implications

I had hoped to be able to discover the counterweight that would best simulate normal double-leg cycling for each participant in the single-leg cycling model. I was then planning on evaluating metabolic costs of similar biomechanics of double-leg and single-leg cycling. Through pilot studies performed in The Neuromuscular Function Laboratory, I was predicting that the previously published studies that indicated the increase in metabolic cost for single-leg cycling (Davies & Sargeant, 1975; Klausen et al., 1982; Stamford, Weltman & Fulco, 1978) over double-leg cycling. In the previous studies it should be noted that no consideration was taken for the fact that a constant workload was evaluated for approximately half the working muscle mass. Once a proper counterweight can be individually applied for single-leg cycling, a re-analysis of metabolic differences and adaptations could be performed.

I feel the metabolic analysis of double-leg cycling compared to counterweighted single-leg cycling needs to be evaluated. If in fact the new counterweighted model of single-leg cycling could produce similar metabolic results at a reduced central demand, this model could possibly provide the ability to reap a greater respiratory capacity. The discussed counterweighted single-leg model also enables the participant to perform this mode for extended period of time without fatiguing, thus exposing the participant to a greater physiological stimulation that would in theory lead to even greater respiratory capacity gains (Gitt et al., 2002; White & Dressendorfer, 2005).

<u>Single-leg Model Additional Thoughts</u>

One final observation in working with this new model of single-leg cycling, I observed that when metabolic costs came out to be similar, RPE was lower thus enabling the participant to perform additional work with little to no extra perceived effort. This was also reported by Abbiss et al. (2010). In addition to the perception of effort, the additional oxygen supply being delivered allowed participants to work longer and at higher percentages of the leg's lactate threshold thus enabling further training stimulation to then lead to increased respiratory capacity (Abiss et al., 2010).

A few studies evaluated an increased oxygen utilization, capillarization, and metabolic capacity in single-leg trained participants, but when the participant performed a global, or double-leg maximal analysis displayed little or no change in total oxygen consumption (VO_{2max}). I suggest this may be due to the fact that although an increased respiratory capacity in each leg is seen individually, perhaps in trained populations the athlete becomes further centrally limited (Abiss et al., 2010; Davies & Sargeant, 1975; Klausen et al., 1982), but in disease populations the adaptations show a globally significant increases in oxygen consumption (Dolmage & Goldstein, 2008). I would anticipate for centrally limited individuals such as CHF or COPD patients an increased oxygen utilization or respiratory capacity could be reached through this single-leg cycling model, but further research needs to be performed to confirm this hypothesis.

Single-leg Model Refined

I have discovered through my research that the counterweighted single-leg cycling model needs to be refined. The ability to determine the proper counterweight for a given individual is vital for the proper application and possible adaptations that may be realized. Further investigations using this model of single-leg cycling should be performed in; refining counterweight selection through a regression equation, metabolically analyzed, and compared to past research on single-leg cycling metabolism compared to double-leg cycling. I would highly recommend that a practice week be implemented for participants to become familiar with the feel of using the counterweight for flexion phase of the crank cycle.

Possible Applications for Model

As discussed in the introduction, I think this single-leg cycling model can serve as a novel approach to evaluate many physiological and training processes. The information obtained in this study created the ground work in exploring a new area of training muscular respiratory capacity. The biomechanical analysis provided evidence that single-leg noncounterweighted cycling is significantly different to normal double-leg cycling. This study also demonstrated a model of single-leg cycling utilizing a counterweight is more similar to normal double-leg cycling than noncounterweighted single-leg cycling. The evaluation of the aforementioned pedal powers will provide information for future investigations on single-leg cycling. Hopefully, this single-leg model provides a very useful way to evaluate biomechanical and physiological parameters in cardiac rehab, physical therapy, pulmonary rehab, and performance.

Table 1: General Demographics of Participants

Sex	Age	Height		Weight		BMI
		CM	IN	KG	LB	
F	50	159.00	62.00	46.00	101.45	18.20
M	40	180.00	70.87	97.00	213.89	29.90
M	40	183.00	72.05	101.00	222.71	30.20
M	45	182.00	71.65	102.00	225.00	30.80
F	27	165.00	64.96	58.00	127.80	21.30
F	30	156.00	61.42	55.00	121.28	22.60
F	40	172.00	67.72	84.00	185.25	28.40
M	35	178.00	70.08	70.00	154.35	22.10
M	35	178.00	70.08	88.50	195.15	27.90
M	45	174.00	68.50	74.00	163.20	24.40
M	45	181.00	71.26	89.50	197.35	27.30
AVG	39	173.45	68.24	78.64	173.40	25.74
STDEV	7	9	4	19	43	4

Table 2:
Means and Standard Deviations for Power Measured Under Different Conditions

Phase	**Condition**	**Mean**	**SD**
Extension	2 Leg	245.7	27.7
	1 Leg - 0 lbs	173.3	23.8
	1 Leg - 20 lbs	201.4	22.3
	1 Leg - 30 lbs	216.8	23.3
Flexion	2 Leg	-85.9	38.0
	1 Leg - 0 lbs	6.7	13.5
	1 Leg - 20 lbs	-38.9	22.8
	1 Leg - 30 lbs	-45.0	23.7

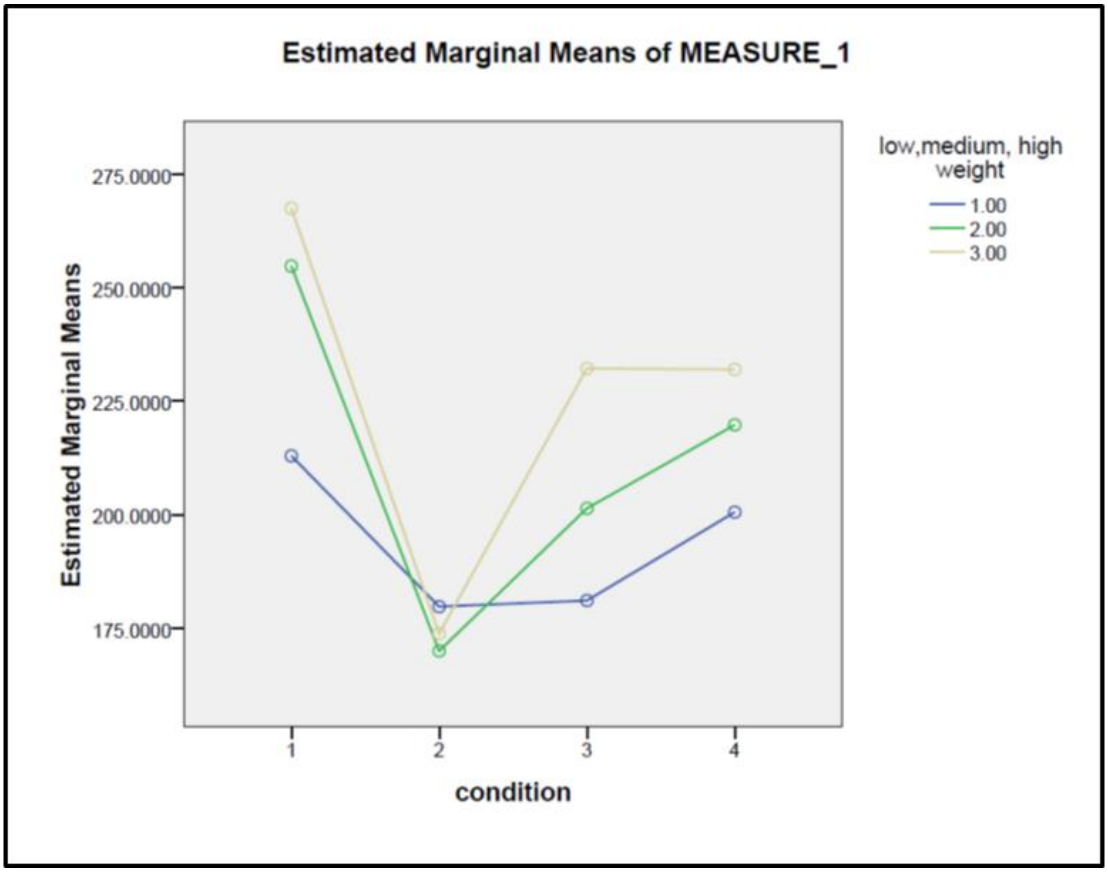


Figure 7: Profile plot for extension using body weight as covariate. Three weights low (1), medium (2) and high (3) were created to show trends in counterweight through the conditions of double-leg (1), single-leg 0 (2), single-leg-20 (3), and single-leg 30 (4).

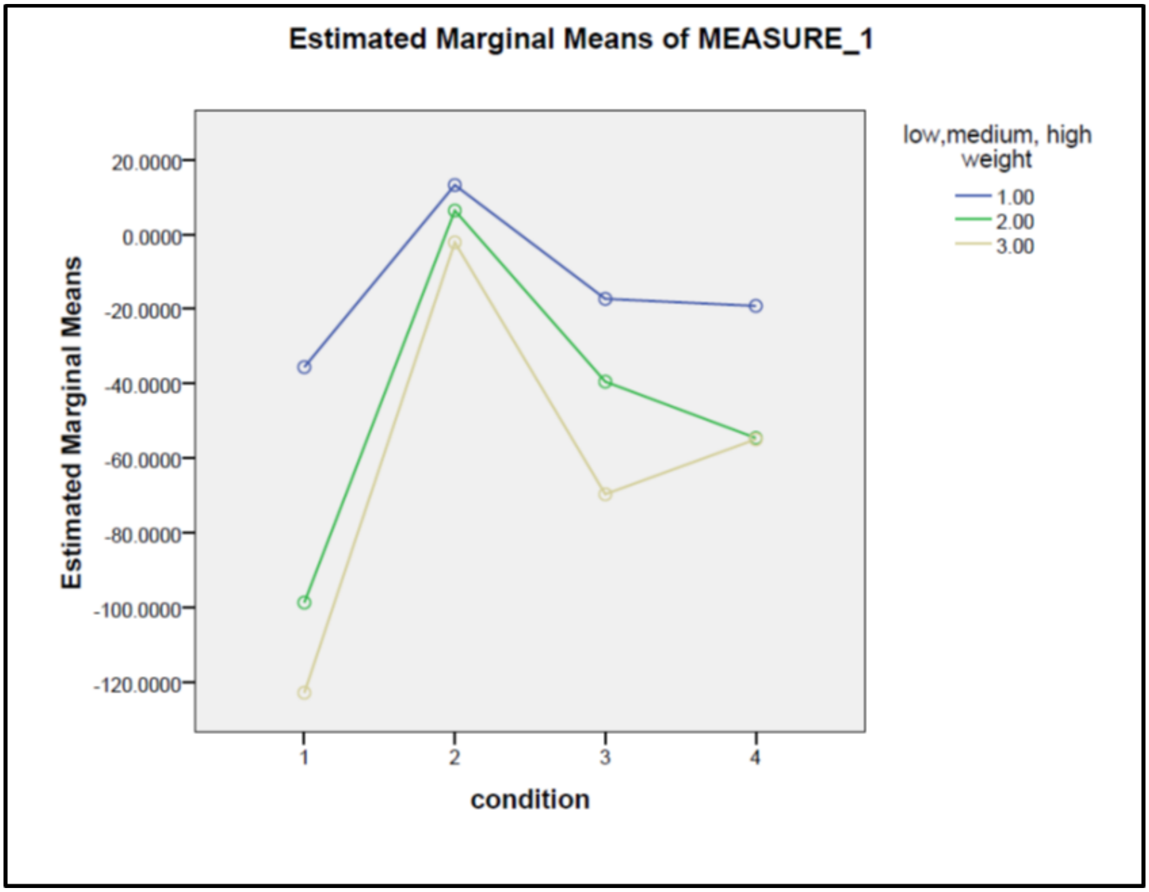


Figure 8: Profile plot for flexion using body weight as covariate. Three weights low (1), medium (2) and high (3) were created to show trends in counterweight through the conditions of double-leg (1), single-leg 0 (2), single-leg-20 (3), and single-leg 30 (4).

CHAPTER 5

SUMMARY, CONCLUSION, AND RECOMMENDATIONS

In this chapter, I summarize the findings of this investigation, draw conclusions, and offer recommendations for future research.

Summary

Biomechanical analysis of pedal powers on double-leg and single-leg cycling reveals very different biomechanical implications. Counterweighted single-leg cycling is a model that can be used to create similar pedal powers in both a single- and double-leg cycling trial. If similar pedal powers can be achieved with a counterweight, implications for rehab, disease states, and performance are vast.

Conclusion

In conclusion, I propose single-leg cycling with no counterweight cannot be thought of as the same task as double-leg cycling. Large biomechanical differences from a normal double-leg cycling model to a noncounterweighted single-leg model are significantly different in both extension and flexion. Based on the information collected from pedal power, we could also make assumptions about other joint kinematics and metabolism that would also most likely be significantly different between the modes.

Counterweighted single-leg cycling provides a model that can be performed assimilating normal double-leg cycling. Unfortunately, due to limited practice sessions and perhaps not experimenting with a wide enough range of counterweights, I was unable to show the single-leg cycling model as not significantly different from normal double-leg cycling. In all conditions, noncounterweighted cycling was clearly substantially more different from the rest of the cycling modes and as the counterweight was increased the pedal powers were more similar to normal double-leg cycling.

Future Recommendations

I strongly recommend that when implementing this type of model for single-leg cycling, especially if the goal is for it to assimilate normal cycling, is to have the participants go through a series of practice sessions. The practice sessions will enable the participants to understand how the counterweight essentially acts as the missing limb, eliminating the need for active flexion and reduced pedal power in extension.

Future recommendations drawn from the collected information in this research would be three fold. First, fine-tune the counterweighted single-leg cycling model and create a regression analysis enabling participant in both clinical/rehab and performance applications to properly select the appropriate counterweight. The proper counterweight selection may differ based on cadence selection and work load being produced. Second, I think further investigation should be performed evaluating the metabolic component of the single-leg model. The ability to demonstrate similar biomechanics and metabolism would solidify that mechanically and physiologically the counterweight model is similar to normal cycling. If the first two recommendations were completed, I believe the final recommendation would be, to re-affirm the previous possible training adaptations and utilize this model in clinical, rehabilitation, and performance groups.

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